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Instruments and Techniques

Testing of an improved ultrasound flowmeter: technical description and results of testing *in vitro*

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Authors' synopsis In vitro *testing was carried out of an advanced model of an intravascular flowgauge based on the ultrasound principle – that is, measurement of the difference in transit time of an upstream and downstream ultrasound pulse. The gauge was tested extensively to determine its sensitivity and stability as well as the influence of changes of haematocrit and temperature on its performance.*

Instruments for the determination of velocity of intravascular flow of blood that are easy to apply are of great interest in medicine (Braunwald, 1965). A number of gauges have been developed which vary widely with respect to the principles of measurement. This paper described further developments of an ultrasonic gauge which was originally described by Plass (1964). This gauge measures the velocity of flow from the difference in upstream and downstream transit time of a periodic sequence of ultrasonic pulses travelling through the blood flow. Scheu, Sager, and Veragut (1965) first showed that such a gauge is capable of measuring the blood flow inside large arteries. However, extensive testing *in vivo* showed that the first designs of the gauge were neither small enough nor mechanically rigid enough to satisfy the demands for practical application and that the electronics had also to be improved to meet more stringent requirements of measurement.

Practical experience suggested that a number of features appeared necessary: (1) overall length of the gauge less than 2·4 cm, diameter 2·3 mm or less, and gauge attachable to a pliable catheter; (2) permissible error of measurement less than ±20%, with linear response over the range of −40 cm/sec to 200 cm/sec; (3) accuracy requirements to be met at deviations of the axis of the gauge from the stream axis up to 15°; (4) calibration of output in mv per cm/s not absolutely necessary.

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Principle of measurement

Two ultrasonic transducers are placed within the stream forming the base and the top of a cylinder whose axis is at an angle α with the stream axis. Let it be assumed that the flow passes through the volume of the cylinder without disturbance. (In reality disturbances occur as discussed below.) The transducers simultaneously emit short bursts of ultrasound during a time shorter than the transit time of sound from one transducer to the other. The emitted pulses are received by the opposite transducers, amplified, and their relative phases compared. When the medium between the transducers is at rest, transit times will be identical. If the emitted signals are in phase the received signals also will be in phase. If the medium flows with a velocity $\vec{v}$, and the unit-vector in the direction of the cylinder-axis is $\vec{n}$ the velocities of the two signals will be $c \pm \vec{v}.\vec{n}$. The signals will have different transit times and this difference is expressed by the formula:

$$\Delta T = \int_0^l \left\{ \frac{1}{c + \vec{v}(x)\vec{n}} - \frac{1}{c - \vec{v}(x)\vec{n}} \right\} dx$$

$$= 2 \int_0^l \frac{\vec{v}(x)\vec{n}}{c^2 - (\vec{v}(x)\vec{n})^2} dx$$

Since $\frac{\vec{v}(x)\vec{n}}{c} \ll 1$ we can write

$$\Delta T = \frac{2}{c^2}\int_0^l \vec{v}(x)\vec{n}\,dx$$

and with $l \cdot \vec{v} \cdot \vec{n} = \int_0^l \vec{v}(x)\vec{n} \cdot dx$

$$\Delta T = \vec{v} \cdot \vec{n} \cdot \frac{2l}{c^2}.$$

If the emitted pulses are in phase, their phase difference at reception will be

$$\Delta\varphi = \Delta T \cdot \omega$$

$$\Delta\varphi = \frac{2l}{c^2} \cdot \omega \cdot \vec{v} \cdot \vec{n}.$$

This difference of phase is converted to a d-c signal by a phase discriminator, the output of which is proportional to the velocity component in the direction of the axis of the gauge.

In a more critical analysis, several conditions have to be fulfilled for the application of this measuring principle within a blood vessel: (1) the velocity of flow between the transducers must be representative of the cross-sectional velocity; (2) for determination of absolute values of flow velocity the angle between the vascular axis and the gauge axis must be known. The error Δv due to axial deviation α between the vessel and the measuring-gauge can be approximated by

$$\frac{\Delta v}{v} = 1 - \cos\alpha,$$

for $\alpha < 15°$ one obtains $\frac{\Delta v}{v} < 0{\cdot}034$;

(3) the phase discrimination should be linear in the whole range of velocities to be measured – that is, in the range of phase differences produced.

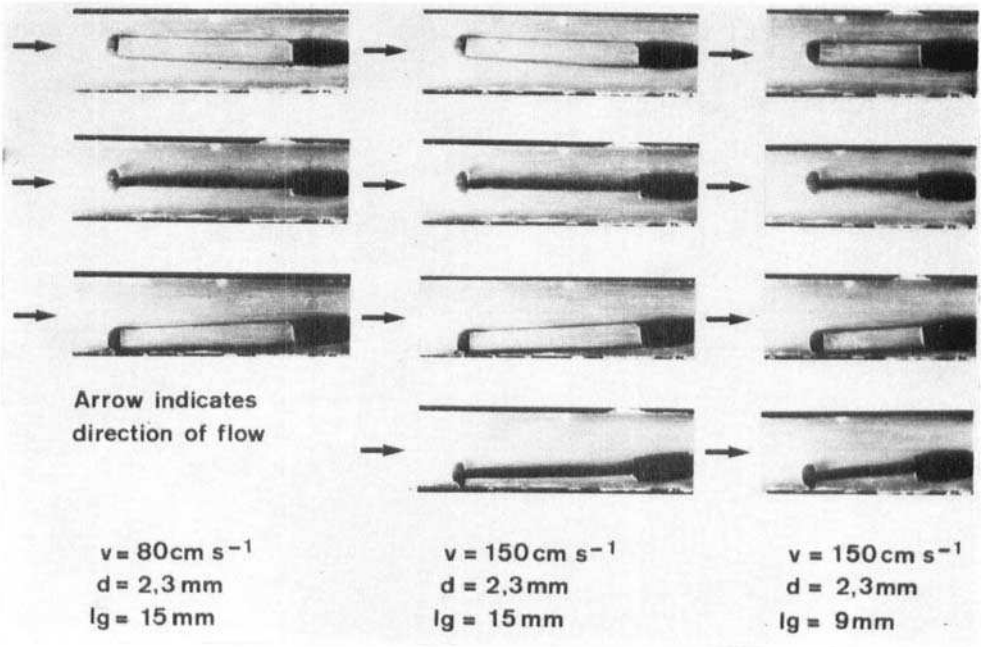
Assuming that the flow is negligibly deformed by the gauge, the measured velocity represents only the flow velocity near or at the gauge. Whether this velocity is representative of cross-sectional velocity depends upon the distribution of flow across the vessel. Theoretically in laminar flow of Newtonian fluids any intravascular device will furnish correct information only if the distance between axis of gauge and axis of flow is $1/\sqrt{2}$ of the vessel's radius.

At the inlet of a large vessel, flow conditions are much more favourable. The velocity distribution in such vessels is relatively uniform so that the velocities representative for the cross-sectional flow occur over a large part of the diameter of the vessels (Kivilis and Reshetnikov 1965a, 1965b).

Description of apparatus

The gauge described as well as the electronic equipment represents the latest model of the ultrasonic flow gauge; it has been developed at the Institute of High Frequency Electronics of the Federal Institute of Technology (Professor F. Borgnis) since 1964. A detailed description will be given by Borgnis *et al.*

FIG. 1 *Flow pattern around two model gauges in a simulated flow channel. At low velocities (column at left) vortex formation is slight. At higher velocities (middle column) the zone of vortices is approx. 20% of the gauges length. With a shorter gauge (column at right) the zone of vortices is proportionally larger. v = flow velocity, d = diameter of gauge model, lg = length of gauge between transducers.*



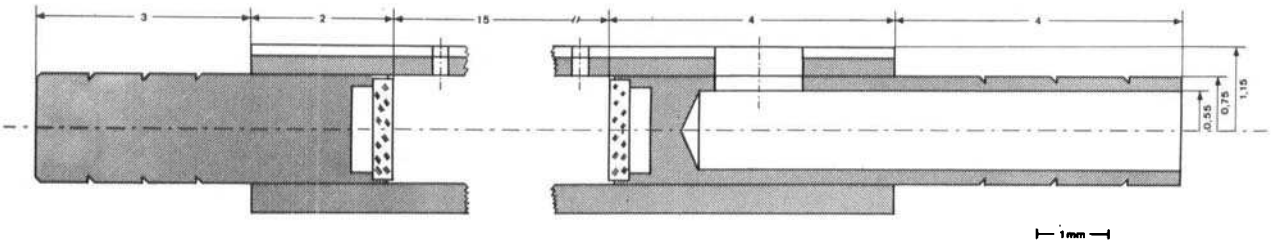
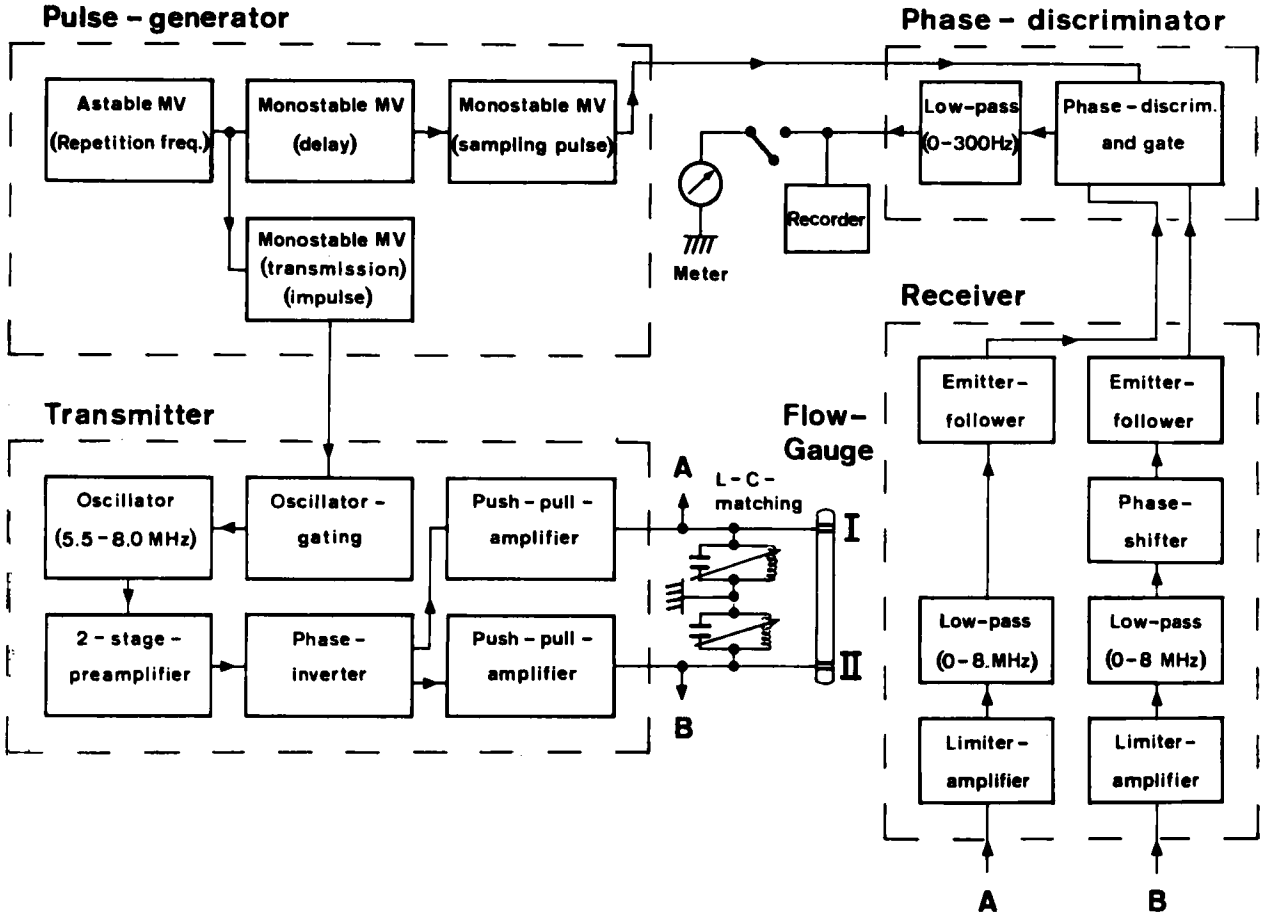


FIG. 2 *Cut-up diagram of the gauge body (see text) (by courtesy of Professor F. Borgnis)*

1 **Measuring gauge** Optimal size and shape of the gauge were studied in a simulated flow channel. To achieve gauge forms that deformed flow as little as possible, experiments were conducted in a plexiglass channel with a square aperture of 1 cm^2. Fine sawdust was added to the water to visualize flow patterns. Knowing the exact exposure time of the Hasselblad camera used, the length of the traces of the particles provided a measure of the velocities in different parts of the flow channel.

Analysis of these photographs (Fig. 1) showed that with a diameter of 2·3 mm, a distance of 15 mm between the transducers was adequate. With such a gauge, vortex formation was negligible at small velocities and even with very fast flow the zone of cortices was small compared with the length of the gauge (about 20%) and was partly compensated by slightly increased speed in the centre of the gauge. With the same diameter, but with a shorter distance between transducers, stream lines are deformed in a way to

FIG. 3 *Block diagram of the apparatus (by courtesy of Professor F. Borgnis).*



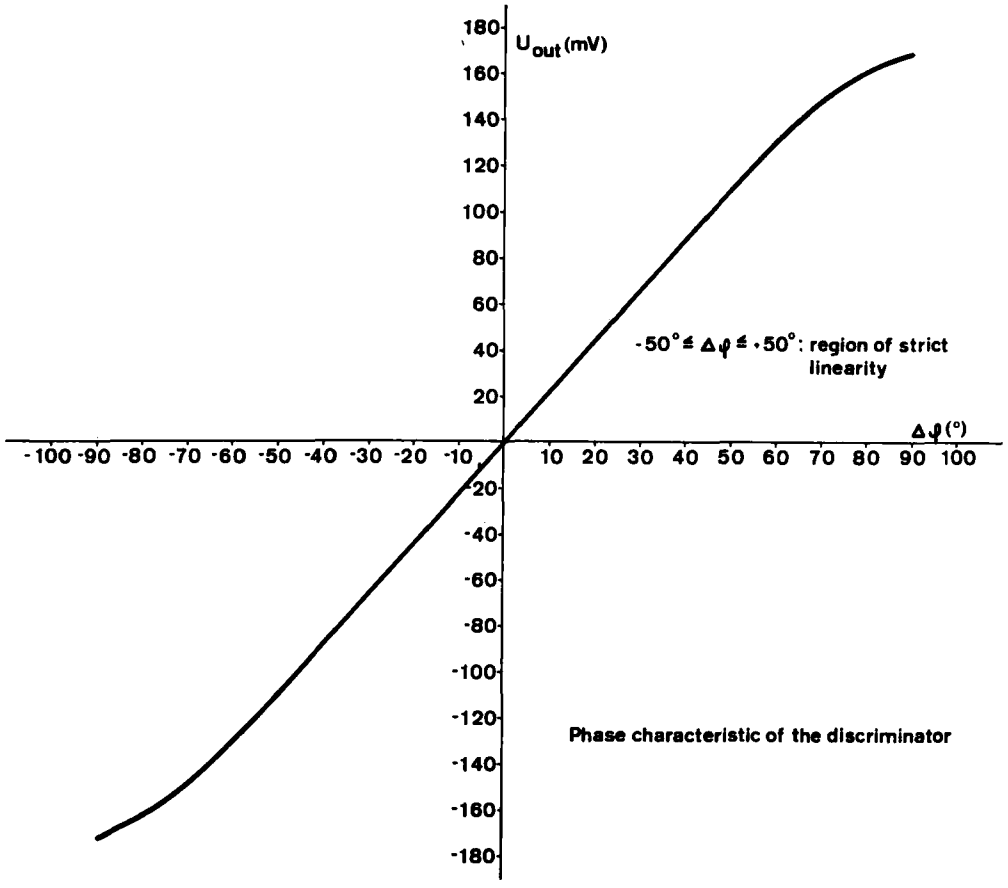
make measurements impossible (Fig. 1). The photographs show also that in our flow channel the velocity was rather uniform except in a narrow zone near the wall of the channel.

2 **Construction of gauge** The gauge body is made of steel tubing with an outer diameter of 2·3 mm and an inner diameter of 1·5 mm, leaving only two narrow strips. The width of these strips represents a compromise between desired mechanical strength and undesirable disturbance of flow, and depends upon the mechanical properties of the material used. A module (diameter 1·5 mm) containing a piezoelectric crystal with a diameter of 1·34 mm is placed at each end of the gauge. The crystals are fixed into a stepped bore hole of 1·4 mm diameter and a depth of 0·2 mm, concentric with the gauge. A second concentric bore hole of 1·2 mm diameter produces an air cushion behind the sound crystal, to prevent sound emission backwards. The ulstrasound transducers (lead zirconate, Sonox 2, Steatit Magnesia, Lauf a.d. Pegnitz, Germany) are then glued on to the remaining portion of the periphery using conductive Araldite (Ciba, Basel). The gauge body is covered with a thin coat of Araldite to prevent mechanical damage and electrical leakage and finally a number 7 F Teflon catheter is drawn over the coaxial copper leads and fixed with Araldite to the gauge body. A detailed description of the gauge will be given by Borgnis *et al.* (to be published).

3 **Electronic equipment** Some electronic data are imposed by the gauge construction: (1) with a sound transducer 0·3 mm thick and with vibration in the thickness mode the resonant frequency is approximately 7 MHz; (2) the transit time of sound for the distance of 15 mm between the transducers is approximately 10 μs, leaving approximately 5 μs for emission.

Figure 3 is the block diagram of the apparatus (Borgnis *et al.* (to be published)). Control is provided by the pulse generators. The oscillator has a high frequency peak-to-peak voltage of 20 V_{pp}. At this stimulation voltage, the received signals have an a-c amplitude of 300 to 800 mV_{pp}, depending on the quality of the gauges used. Since the two receiving channels and the two transducers are not completely symmetrical, a phase shifting stage was incorporated into the receiving part of

FIG. 4 *Phase characteristic of the discriminator.*



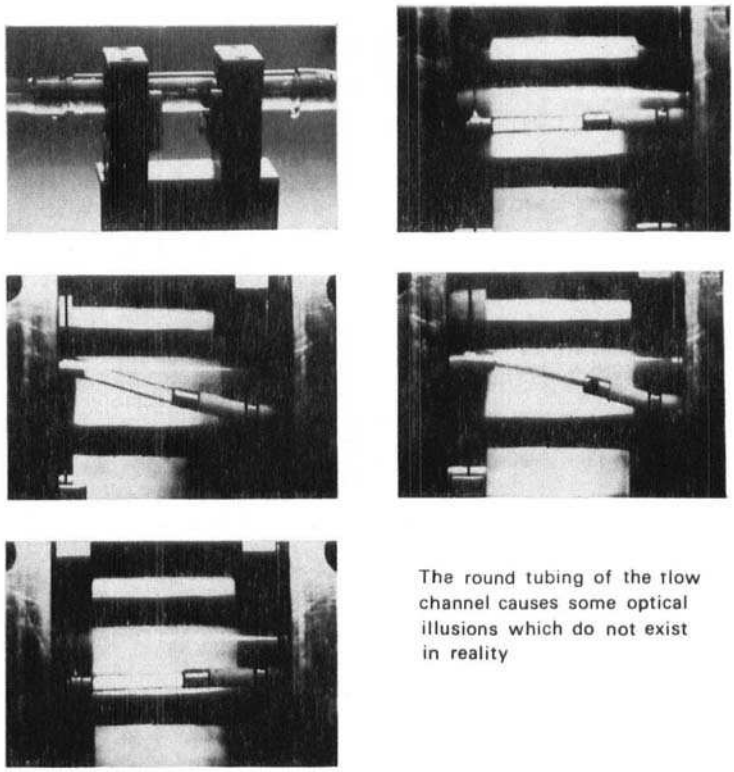


FIG. 5 *Flow channel for testing the gauges in different positions. The gauge can be fixed in various positions by means of slide wires. Optical distortion caused by the round tubing gives the impression of curved gauges. Upper left: the flow channel assembly. Upper right: the gauge in central position. Fixation by the slide wires is visible at both ends of the gauge. Middle row: the gauge fixed in oblique positon and rotated by 90°. Lower left: the gauge positioned near the wall of the flow channel. Recordings were also made with the gauge rotated by 90°.*

the instrument to compensate for the constant difference of phase of the two signals.

Special attention was given to assure that the phase discriminator functioned linearly over a wide range. Figure 4 shows the characteristics of the discriminator. With a 7 MHz signal, a gauge 15 mm long and a velocity of flow of 150 cm/sec the shift of phase produced in the medium water is approximately 50°. Some deviations from linearity are to be expected above this velocity; they do not exceed 15% up to velocities of 180 cm/sec. The influence of this unlinearity is discussed below.

In vitro *testing procedure*

Figure 5 shows the flow channel developed for the testing of the gauges in every desired position. Two roller pumps in parallel provide flows up to 200 cm/sec. Since the flow is slightly pulsatile the signals are damped using appropriate resistive-capacitive filters and recorded on a x-t recorder. The temperature is held constant within a range of ±0·5°C using a thermostat and two heat exchangers; the temperature can be varied between 0° and 80°C. The output of the pumps as a function of speed is determined initially with a stopwatch and a graduate cylinder.

Results

All measurements were performed in at least duplicate with water and outdated human blood. The gauge was tested as a function of temperature, haematocrit, and gauge position with respect to the stream axis. Water at 20° C was used as a standard for continuing control of properties of the gauges.

1 **Gauge testing in water** Measurements in water were performed to provide a standard for long-term experiments. Sensitivity was arbitrarily set at 0·2 mV per cm/sec. Over a period of 8 hr, the sensitivity varied by ±5% and differed from one gauge to the other. The output of the instrument was linear within ±10% from 0 cm/sec to 200 cm/sec (Fig. 6). No change of sensitivity with temperatures between 20° and 40°C could be detected.

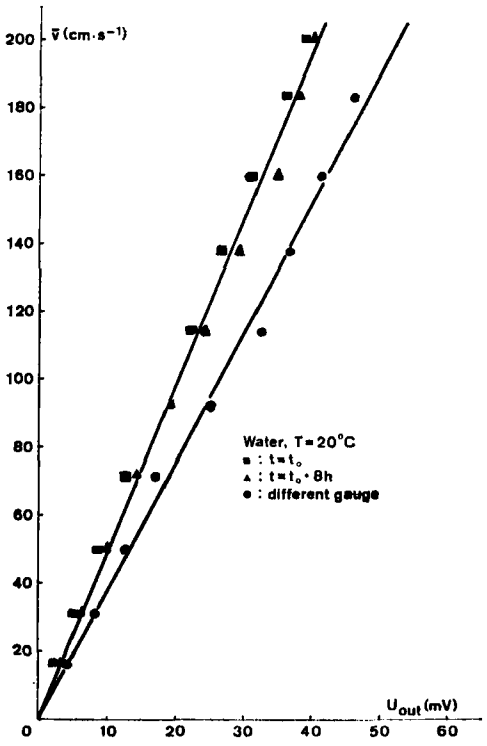


FIG. 6 *Results of testing two gauges in water of 20°C. Sensitivity did not change during one eight-hour experiment. Sensitivity of the gauges in water at 20°C could therefore be used as a standard for long-term experiments. Different gauges show different sensitivities.*

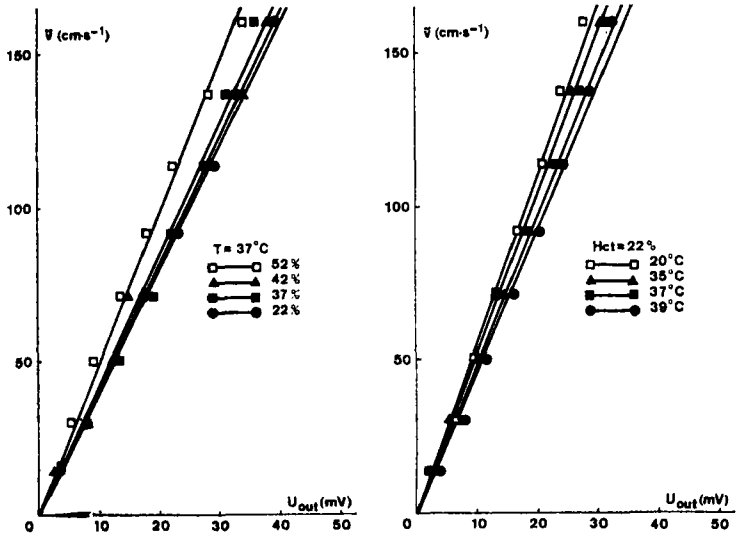


FIG. 7 *Output of the gauge with different haematocrits (left) and temperatures (right) (see text). The heavy lines represent the arithmetic means through all measuring points of an experiment.*

2 **Gauge testing in human blood** Haematocrit was controlled before and after each run. When haemolysis set in, the blood was discarded.

a *Effects of changes in haematocrit* Measurements were performed at haematocrit levels from 22 to 52%, achieved by mixing erythrocytes and plasma in various proportions. Linearity of output was not affected by the different haematocrit levels and prevailed up to velocities of 160 cm/sec. Only with very high haematocrits and very high velocity were slight deviations from linearity observed (Fig. 7). Figure 8 summarizes the changes of sensitivity with increase of haematocrits.

b *Effects of changes in temperature* All measurements with blood were performed at 20°C, 35°C, 37°C, and 39°C. No change of linearity was observed over this range (Fig. 7). As shown in Fig. 8 sensitivity increased slightly with rising temperatures.

c *Effects of changing position of the gauge* This group of experiments was of special importance since *in vivo* strictly axial position can hardly be controlled. The gauge was first positioned at an angle of 15° to the axis of the channel and so that both metal strips were in the same plane as the central axis of the channel. Additional tests were conducted with the gauge in the same position but reorientated by rotation of 90° about its own axis. We also examined the influence of positioning the gauge as close to the walls as possible (Fig. 5). These experiments were performed at a haematocrit of 40% and a temperature of 37°C (Fig. 9). When the strips were coplanar with the axis of flow in the oblique position (pos. B), sensitivity was identical with that of a centrally positioned gauge, as drawn in for comparison (pos. A). When the gauge was rotated 90° (pos. C), sensitivity was somewhat higher and reached 0·25 mV per cm/sec instead of the previous 0·20 mV per cm/sec. When the gauge was placed in a lateral position (pos. D), results became considerably worse: scatter of the individual measurements increased, sensitivity seemed to be decreased, and linearity could not be determined.

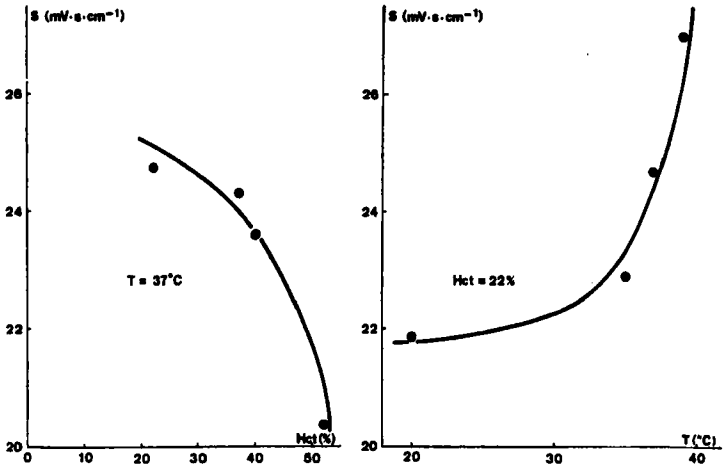


FIG. 8 *Sensitivity of the gauge as a function of haematocrit (left) and temperature (right). For clarity the abscissa was chosen at a greatly enlarged scale.*

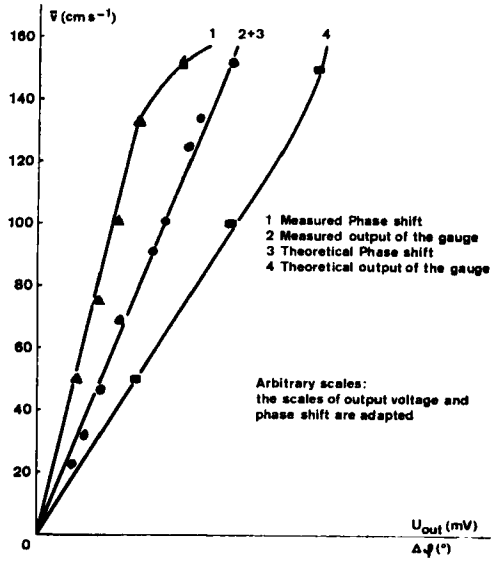
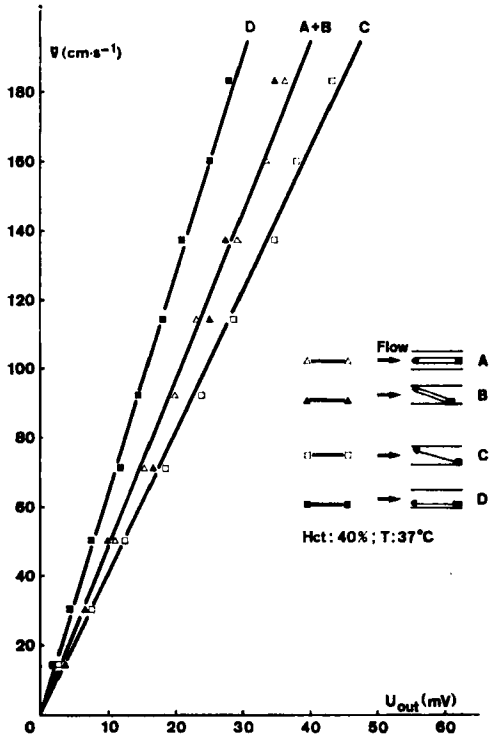


FIG. 10 *Difference of theoretically expected phase shift with rising velocities and observed measurements of the gauge (see text).*

FIG. 9 *Output of the gauge with different positions within the flow channel. The positions sketched at the right of the graph correspond to the photographs in Fig. 5.*



Discussion

This new ultrasound gauge fulfils the requirements of the biologist for a gauge that can be easily introduced into a large vessel. Its size and mechanical properties allow placement into the pulmonary artery of dog or even man (Fricke, Studer, and Scheu, 1970). Furthermore, the vigorous testing *in vitro* showed that accuracy requirements were fulfilled. Linearity is satisfying (better than ±10% up to 200 cm/sec). Only with very high haematocrits and very high velocities or when the gauge is positioned close to the walls are some deviations from linearity observed. Since the response of the gauge is directionally sensitive, measurements can be regarded as linear within the limits required by the biologists. With position in midstream and with angles of up to 15° between gauge and axis of flow the errors of measurement are less than 20% while the linearity is not affected. At the moment the sensitivity of the gauge can not be calibrated *in vitro*. It is necessary for this purpose to use blood at about the temperature and haematocrit of the experimental animal. A closed circuit pump with a small priming volume and temperature control is under development.

One further point needs discussion. We have shown that our phase discriminator trespasses its linear range at high flow velocities. The complete instrument on the

other hand does not show this same deviation. From observations of the flow in the gauge we must conclude that the fast-varying eddies and vortices between the two transducers lead to an increased non-linear rise in phase-shift at high velocities. This effect of the output would be compensated by the nearly inverse behaviour of the phase discriminator. Figure 10 shows the theoretical output of the flowmeter assuming linear relation between phaseshift and velocity and taking into account the effective shape of the phase discriminator. Also the measured output is plotted, which demonstrates the expanded linear range. Thus the new ultrasonic intravascular flowmeter furnishes reproducibly linear measurements of satisfactory reliability.

Summary

A new model of an ultrasonic catheter flowmeter is small and stable enough to allow measurements in the pulmonary artery of dogs and even man. The output is linear between −40 cm/sec and 180 cm/sec. Rising temperatures increase and rising haematocrits decrease the sensitivity to a minor extent. When the axis of the gauge is at an angle to the axis of flow no significant changes of sensitivity occur up to angles of 15°.

The work reported here would not have been possible without the help of Professor F. Borgnis (Institute of High Frequency Electronics of the Federal Institute of Technology, Zurich) and his collaborators. We gratefully acknowledge the help of F. Brunner, P. Fruttiger, G. Horak, F. Kail, H. Peyer and A. Schmid, and of K. G. Plass who originated the physical principles. We also greatly appreciate the help with velocity measurement given to us by the Institute of Photography of the ETH, and the capable help of Mr. R. Danieli in constructing the gauge.

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